Application of Linux Audio in Hearing Aid Research

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Abstract

Development of algorithms for digital hearing aid processing includes many steps from the first algorithmic idea and implementation up to field tests with hearing impaired patients. Each of these steps has its own requirements towards the development environment. However, a common platform throughout the whole development process is desirable. This paper gives an overview of the application of Linux Audio in the hearing aid algorithm development process. The performance of portable hardware in terms of delay, battery runtime and processing power is investigated.

Keywords

Hearing aids, algorithm development, low delay, feedback cancellation

1 Introduction

The introduction of the first commercial hearing aid with digital signal processing in 1996 substantially changed the methods used in hearing aid research. Whereas typical analogue hearing aids have signal processing blocks for frequency shaping, dynamic compression and static notch filters for feedback cancellation, hearing aids with digital signal processing allow the implementation of algorithms which have no counterpart in the analogue domain. Another recent development in hearing aid technology is the availability of a low-delay wireless binaural link between the left and right ear in commercial hearing aids. This link opens up new categories of signal processing: Binaural signal analysis and presentation of a binaurally enhanced audio signal.

Hearing aid algorithms have to be beneficial for the hearing impaired in an objectively accessible way. The most important benefit is an improvement of speech intelligibility in noisy environments. Also less obvious improvements, such as reduction of listening effort, increase of listening comfort, improvement of spatial perception, and artifact reduction are important. The still limited processing power and a low battery capacity make it necessary to avoid every unneeded processing cycle. Thus, hearing aid algorithms are substantially tested in as many and as realistic situations as possible. A solution for field testing is the implementation of a hearing aid in a programmable and portable signal processing system, e.g., the DASi [1]. The drawback of such DSP-based systems is the need of implementing the typically complex algorithms in low level languages, which requires large effort for re-configuration and modification. This is different if standard hardware and standard operating systems can be used for algorithm development and evaluation. Recent developments in Linux Audio suggest that the Linux platform might be well suited for hearing aid research. A key feature is the easy access to low delay real-time audio processing on standard hardware.

In the first part of this paper, an overview of algorithms for digital hearing aids is given. Evaluation methods are briefly described. In the second part, the application of Linux Audio for hearing aid research is demonstrated by the example of a portable field testing device. Processor performance data and the analogue delay of a Linux based hearing aid is given in the results section.

2 Overview of Algorithms for Digital Hearing Aids

Key algorithms of hearing aids – be it digital or analogue – is frequency shaping and dynamic compression, for correction of audibility and loudness recruitment. Loudness recruitment denotes the reduced dynamic range between hearing threshold and uncomfortable loudness level often found in sensorineural hearing loss, the most common type of hearing loss. For an increase of signal-to-noise ratio (SNR) and thus a better understanding in adverse listening conditions, hearing aids often use mul-

tiple microphones for directional filtering. The easiest approach are delay-and-sum beamformers with a fixed beam, which can reach up to 6 dB gain in SNR for on-axis sounds. More elaborated are adaptive beamformers [2] and sidelope cancellers. Single channel noise reduction [3] and binaural de-reverberation [4] are algorithms which can increase the listening comfort and reduce the listening effort. However, speech intelligibility can not be improved significantly by this class of algorithms [5]. Hearing aids also introduce artifacts, which can reduce audio quality or even speech intelligibility. One of the largest problems in hearing aids is the feedback howling, caused by leakage of the receiver (speaker) signal back to the microphones, and a typically large amplification. Feedback artifacts can be reduced by three categories of algorithms: Adaptive feedback cancellation schemes [6] estimate the feedback signal in an adaptive filter and remove the estimated feedback signal from the microphone input. Binaural information can be utilised to increase stability of hearing aids [7], and feedback problems can be avoided by opening the loop in critical frequency ranges with non-linear operations in the hearing aid, e.g., frequency transposition or phase distortion [8]. For a more detailed overview of algorithms see [9] and [10].

2.1 Evaluation methods

For the evaluation of hearing aid algorithms both objective evaluation methods and subjective methods are necessary. The improvement of SNR is a canonical measure for algorithms which aim to remove or attenuate noise, e.g., directional microphones and noise reduction schemes. The Speech Intelligibility Index (SII) is commonly used to predict speech intelligibility in non-fluctuating noise [11]. Extensions of the SII include the addition of binaural processing [12]. Quality measures range from technical measures like spectral distortion [13] to methods based on audio perception models [14].

To measure speech intelligibility subjectively, typically the speech recognition threshold (SRT) is measured by adaptively changing the level of speech or the SNR of speech in noise until a certain percentage of the presented speech signal can be understood [15]. Audio quality and other subjective attributes can be assessed by quality scaling or paired comparison.

For the subjective evaluation of directional filtering and feedback cancellation schemes, a real-time version of the algorithms under test is necessary, to account for head and jaw movements. Although most of the evaluation methods can only be applied in laboratories, an evaluation of the algorithms under test in real-life conditions or even in the field are desirable. However, the more realistic the test conditions, the less reproducible are the results.

2.2 Feedback howling and low delay constraints

The amplification of hearing aids is limited by the feedback from hearing aid receivers to the microphones. The criterion for feedback howling to occur is fulfilled if the amplification of the hearing aid is larger then the attenuation of the feedback path – mainly the damping of the ear mould, and if the round trip phase, i.e., hearing aid and feedback path, is an integer multiple of 2π. In Figure 1 the amplitude and phase response of a hearing aid feedback path is plotted. Frequencies at which the criterion is fulfilled are marked by a circle. The feedback criterion is time dependent because of changes in the feedback path, e.g., ear mould leakage by jaw movements, reflections by phone receivers and room acoustics, and by the time variant amplification of the hearing aid, caused by dynamic compression and signal enhancement. The number of frequencies at which the criterion is fulfilled – and thus feedback can possibly occur – increases with the delay (see phase responses for delays of 1 ms and 5 ms in Figure 1).

3 Implementation of a Linux-based field test hearing aid

For subjective evaluation of advanced hearing aid algorithms with hearing impaired subjects, a portable field test hearing aid based on a Netbook computer with a dedicated USB audio device to connect hearing aid shells has been implemented. This hardware runs UbuntuStudio with a real-time patched Linux 2.6.24 kernel, generic ALSA driver for USB sound cards, and the JACK audio connection kit. The Hörtech Master Hearing Aid (MHA, [16]) connects to JACK and hosts the hearing aid algorithms. The MHA can be fitted to an individual hearing loss using a TCP/IP configuration interface. End user control (i.e., program switching, volume control) and data inspection and logging is also possible via the network connection. A picture of the system is shown in Figure 2.

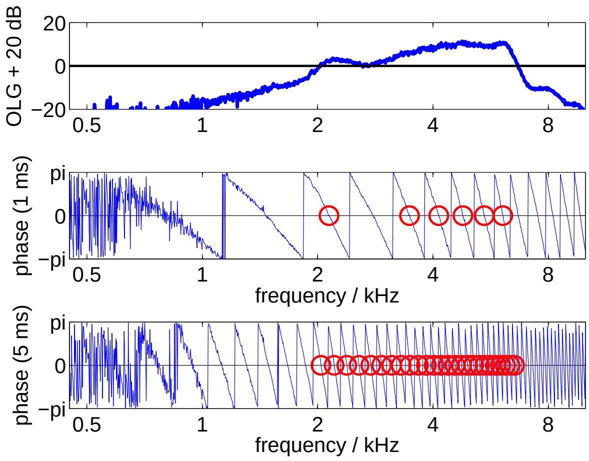

Figure 1: Amplitude (top panel) and phase response (middle and bottom panel) of a hearing aid feedback path. Circles mark the frequencies at which the feedback criterion is fulfilled. The phase response is given for a hearing aid with 1 ms delay (middle panel) and with 5 ms delay (bottom panel).

Figure 2: Portable field test hearing aid. The host PC is an Asus Eee PC 701 running UbuntuStudio 7.10 and the Master Hearing Aid [16]. A dedicated USB audio device is used for connecting the hearing aid shells to the computer.

The sound card[1] provides four input channels and two output channels, and is USB powered. Hearing aid shells (silver headsets in Figure 2) with microphones can be directly connected via cables. With a low output impedance of 7.6 Ω it is able to drive low-impedance hearing aid receivers (speakers) and provide sufficient output level for subjects with a large hearing loss.

3.1 CPU and battery performance

During the development of the field test hearing aid, two different Netbook computers have been tested and compared regarding their processing performance and battery runtime. The data are shown in Figure 3 and Figure 4. A software hearing aid with four different programs was running during the test. The software hearing aid was running at a JACK sampling rate of 32 kHz, and with 1 ms blocks. Internally, the signal was resampled to 16 kHz. The processing contained level adaptation and frequency equalisation, four alternative types of signal enhancement (directional filtering, two versions of single channel noise reduction and binaural coherence based noise reduction, see [17] for details on the algorithms). The signal enhancement was followed by a multi-band dynamic compression scheme, which was fitted to a sample hearing loss (typical high-frequency hearing loss). The compressor, coherence filter and one single channel noise reduction are based on an overlap add FFT method [18].

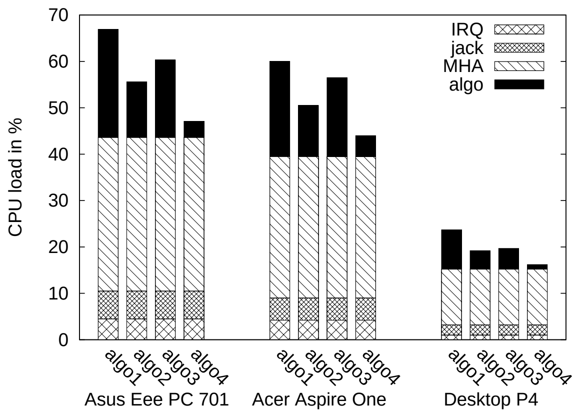

Figure 3: Performance of a field test hearing aid system on different hardwares. In addition to the algorithm CPU time, the CPU time used by the jackd sound server and the sound card interrupt handler was measured.

3.2 Acoustic Delay of PC-based hearing aids

The acoustic delay in hearing aids should be as small as possible. A maximum acceptable delay of 20 ms was found for hearing aids with closed fitting (the ear canal is closed by the ear mould) [19]. For hearing aids with open fittings, i.e., an

[1]The sound card was developed for application in a field test hearing aid by OFFIS e.V., Oldenburg, Germany.

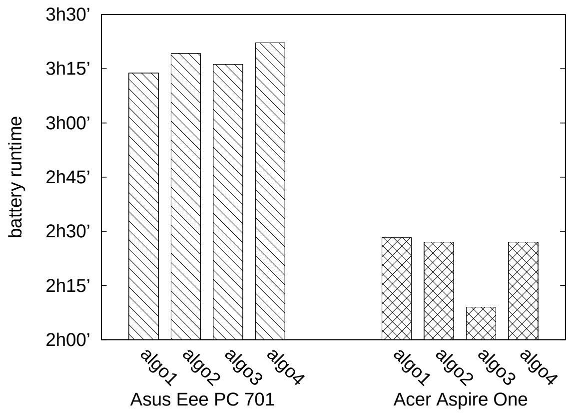

Figure 4: Runtime of the hearing aid prototype with a fully charged battery and the lid closed.

ear mould with a vent or only a slim tube with a dome, the delay need to be below approximately 6 ms to avoid disturbances [20]. The delay of typical audio signal processing algorithms with block processing is determined by three sources of delay: (i) By the group delay of the algorithm itself, which can be frequency dependent and time variant, (ii) by block processing, which is typically two times the period size, and (iii) by the anti-aliasing filters and signal handling in the audio interface and AD- and DA-converters.

The total delay of a hearing aid with no algorithmic group delay (i.e., identity processing), was measured using JACK. The total delay τ_t is the delay from the electrical input to the electrical output. To allow an easier measurement, instead of opening the loop at the electric in- and output, a direct analogue connection from output to input was made and the loop was opened digitally, i.e., the delay from a JACK client output to a JACK client input via the hardware was measured. For a validation of the results, all measurements have been repeated, drop-outs have been monitored and the total harmonic distortion (THD) has been measured to ensure a non-distorted signal output.[2] The data are shown in Table 1. The shortest delay of 1.77 ms is reached with the Echo Layla 3G at 100 kHz sampling rate and 32 samples period size. However, hearing aids are typically processed at low sampling rates between 16 and 32 kHz. A downsampling from 100 kHz to the desired hearing aid sampling rate within the software requires anti-aliasing filters as well, which will introduce an additional group delay. The delay of the hardware anti-aliasing filters and signal handling τ_{sc} has been estimated by subtracting number of periods times the period size from the total delay.[3] The delay τ_{sc} contains the delay of the anti-aliasing filter of both, AD- and DA-converters.

Device	f_s/kHz	P	τ_{sc}/ms	τ_t/ms
Echo	32	32	2.81	4.81
Layla 3G	44.1	32	2.04	4.22
	44.1	**64**	**2.04**	**4.94**
	48	32	1.88	3.88
	64	32	1.27	2.77
	88.2	32	0.92	2.01
	96	32	0.84	1.84
	100	32	0.81	1.77
RME	32	64	3.34	7.34
HDSP9652 +	**44.1**	**64**	**2.68**	**5.58**
Behringer	48	64	2.52	5.19
Ultramatch	64	64	2.16	4.16
SRC2496	64	128	2.14	6.14
	88.2	128	1.8	4.71
	96	128	1.73	4.40
RME	32	64	2.13	6.12
HDSP9652 +	**44.1**	**64**	**1.61**	**4.51**
Behringer	48	64	1.46	4.12
Ultragain				
ADA8000				
RME	32	64	2.09	6.09
HDSP9632+	**44.1**	**64**	**1.52**	**4.42**
ADI8QS	48	64	1.4	4.06
	64	128	1.55	5.55
	88.2	128	1.12	4.02
	96	128	1.03	3.7
OFFIS	16	16	6.81	9.81
USB SC-4/2	32	32	4.38	7.38
	44.1	**64**	**4.08**	**8.44**
	48	64	3.96	7.96
	96	128	1.97	5.97

Table 1: Total delay τ_t from electrical input to electrical output, including delay caused by block processing and the delay τ_{sc} caused by digital transmission and anti-aliasing filters, for several professional audio devices clocked at the sampling rate f_s and using a JACK period size P.

[2] The measurement of THD was necessary because some sound cards produced audible distortion at short period sizes, but JACK did not report an error or xrun.

[3] For period sizes below 1 ms the Echo Layla 3G driver seems to use an addition third period for buffering, which adds to the delay.

4 Discussion

Recent developments in the Linux Audio world made it possible to develop and evaluate algorithms for digital hearing aids on the Linux operating system. A portable field test hearing aid with a low overall delay can be implemented based on standard hardware and the Linux operating system.

4.1 Hardware and performance

The processing performance of miniature laptop PCs is sufficient for many advanced hearing aid algorithms. However, the runtime without recharging the battery is below four hours, which is not sufficient for full-day testing. The round-trip delay of most tested sound cards is in the order of 5 ms at 44.1 kHz. The tested USB sound card has a delay of 8.4 ms, however, this is the only portable bus powered sound card with a sufficient number of input channels, which is required for multi-microphone processing, e.g., beamforming.

The advantage of dedicated DSP-based field testing environments is the efficient execution of algorithms and low-delay capabilities. However, to achieve this benefit, low-level program code has to be written, which is typically time consuming and does not offer access to high-level libraries.

4.2 Usability and Accessibility

A major advantage of using Linux Audio against DSP based hardware solutions is the *accessibility* of low-delay audio performance and a development environment: Real-time patched kernels are available from several audio-related Linux distributions. Most standard hardware is supported by these mainstream kernels. Sound cards and to other sources and sinks of audio are easily accessible via ALSA and the Jack Audio Connection Kit. This is opposed by a restricted access to hardware and development environments for DSP-based hardware solutions.

4.3 Licensing

Application in industrial context often requires closed-source software development. Suggesting Linux as a development platform sometimes meets with reservations from industry decision makers who fear that the open source licensing of the platform might somehow spread to their product. The past has also seen court sentences against companies who failed to comply with the conditions of the GPL software that they distributed. Our point of view is that it is possible to develop closed software on the Linux platform without infringing any licenses, and this does not place a greater burden on the development company than using closed-source software. Instead, it opens a wealth of existing third-party libraries that may help in getting the product to market fast. This requires careful consideration of what software components to use, how they are being used, knowledge of the relevant licenses, and compliance with their terms. Sometimes this means, that a third-party library can be used, but may only be linked dynamically and shipped together with its source code. Other free software components may not be used at all, or different terms of licensing have to be negotiated with the copyright holder. This process of considering software components and their licenses, however, does not differ between closed and open software development platforms.

5 Conclusions

Based on the results of this study and the observation of general aspects of Linux Audio it can be concluded that Linux Audio is applicable in hearing aid research. Requirements of technical kind, e.g., processing performance and audio delay are fulfilled, and the availability of software and supported hardware is sufficient for the research work. The authors think that these features make it superior to DSP-based hardware solutions.

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References

[1] U. Rass and G. H. Steeger. A high performance pocket-size system for evaluations in acoustic signal processing. *EURASIP Journal of Applied Signal Processing*, 3:163–168, 2001.

[2] L. J. Griffiths and C. W. Jim. An alternative approach to linearly constrained adaptive beamforming. *IEEE Trans. on Antennas Propagation*, 30:27–34, 1982.

[3] Y. Ephraim and D. Malah. Speech enhancement using a minimum mean-square error short-time spectral amplitude estimator. *IEEE Trans. ASSP*, (32):1109–1121, 1984.

[4] Th. Wittkop and V. Hohmann. Strategy-selective noise reduction for binaural digital hearing aids. *Speech Communication*, 39:111–138, 2003.

[5] M. Marzinzik. *Noise reduction schemes for digital hearing aids and their use for hearing impaired*. PhD thesis, University of Oldenburg, 2000.

[6] J. M. Kates. Feedback cancellation in hearing aids: Results from a computer simulation. *IEEE Trans. Signal Process.*, 39:553–562, 1991.

[7] V. Hohmann, V. Hamacher, I. Holube, B. Kollmeier, and T. Wittkop. Method for the operation of a hearing aid device or hearing device system as well as hearing aid device or hearing device system, March 2006. US Patent 7,013,015.

[8] M. Schroeder. Improvement of acoustic-feedback stability by frequency shifting. *Journal of the Acoustical Society of America*, 36(9):1718–1724, 1964.

[9] J. M. Kates. *Applications of digital signal processing to audio and acoustics*, chapter Signal processing for Hearing Aids. Kluwer Academic Publishers, 1998.

[10] H. Dillon. *Hearing Aids*. Boomerang Press, Australia, 2001.

[11] ANSI-S3.5. *American national standard methods for the calculation of the speech intelligibility index*. American National Standards Institute, New York, 1997.

[12] R. Beutelmann and T. Brand. Prediction of speech intelligibility in spatial noise and reverberation for normal-hearing and hearing-impaired listeners. *Journal of the Acoustical Society of America*, 120(1):331–342, 2006.

[13] K. Eneman, A. Leijon, S. Doclo, A. Spriet, M. Moonen, and J. Wouters. *Advances in Digital Speech Transmission*, chapter Auditory-Profile-Based Physical Evaluation of Multi-Microphone Noise Reduction Techniques in Hearing Instruments. John Wiley and Sons, Ltd, 2008.

[14] R. Huber and B. Kollmeier. Pemo-q. a new method for obejctive audio assessment using a model of auditory perception. *IEEE Transactions on Audio, Speech and Language Processing*, 14:1902–1911, 2006.

[15] K. Wagener and T. Brand. Sentence intelligibility in noise for listeners with normal hearing and hearing impairment: influence of measurement procedure and masking parameters. *International Journal of Audiology*, 44(3):144–156, 2005.

[16] G. Grimm, T. Herzke, D. Berg, and V. Hohmann. The Master Hearing Aid – a PC-based platform for algorithm development and evaluation. *Acta Acustica united with Acustica*, 92:618–628, 2006.

[17] K. Eneman, H. Luts, J. Wouters, M. Büchler, N. Dillier, W. Dreschler, M. Froehlich, G. Grimm, V. Hohmann, R. Houben, A. Leijon, A. Lombard, D. Mauler, M. Moonen, H. Puder, M. Schulte, A. Spriet, and M. Vormann. Evaluation of signal enhancement algorithms for hearing instruments. In *Proceedings of the 16th European Signal Processing Conference*, Lausanne, Switzerland, 2008.

[18] J. B. Allen. Short term spectral analysis, synthesis, and modification by discrete Fourier transform. *IEEE Trans. Acoust., Speech, Signal Processing*, 25(3):235–238, June 1977.

[19] M. A. Stone and B. C. Moore. Tolerable hearing aid delays. i. estimation of limits imposed by the auditory path alone using simulated hearing losses. *Ear and Hearing*, 20(3):182–192, 1999.

[20] M. A. Stone, B. C. Moore, K. Meisenbacher, and R. P. Derleth. Tolerable hearing aid delays. v. estimation of limits for open canal fittings. *Ear and Hearing*, 29(4):601–617, 2008.